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IMPACT ACCELERATIONS

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## Introduction

Impact acceleration may be encountered during normal as well as emergency phases of spacecraft operations. The impact loads experienced during normal flight phases occur primarily on landing of the spacecraft upon its return to earth. Recovery systems used or considered for use in spacecraft designs have included conventional single and multiple parachute canopies, retrorockets, inflated fabric spheres, and parawings or parasails. Within existing technology and primarily weight limitations it has not been practical to allow descent and horizontal drift rates to be adequately controlled within the range of impact velocities that would not be hazardous to the crew under all adverse circumstances. Exact knowledge of the physical environment to which the astronauts might be exposed with a particular spacecraft and its recovery system for all potential environmental variables, that is, impact surface, wind, impact angle, etc., is absolutely essential for realistic risk analysis and evaluation of protective requirements.

The severity of the impact experienced during spacecraft landing can be reduced considerably by controlling the site of the landing. By such control the impact surface and wind conditions that are most favorable may be selected. Water, or flat, soft terrain have generally proven to produce less severe impacts. Data are available on the dynamics of water impact (17) and, more specifically, on the water and land impact characteristics of the Apollo spacecraft (6, 93). The descent rate at impact may range up to 8.5 m/sec if the recovery system deploys properly. In a design such as the Apollo spacecraft where

three recovery parachutes were used, the descent velocity could be as high as 15.2 m/sec. The resulting impact pulses that occur even under nominal conditions are typically high amplitude, short rise time accelerations as shown in figure 1.

If a catastrophic failure occurs on the launch pad during the final portions of the preflight preparations, short duration, high amplitude acceleration may be required to catapult the space vehicle crewman safely away from the launch vehicle. This same emergency escape system may be required during the initial phase of launch vehicle acceleration if there is failure of the propulsion or guidance systems. The acceleration environment associated with the use of the escape system is more complex as the launch vehicle achieves higher velocities while it is still within the earth's atmosphere. This more complex environment is due to interaction with the windstream and rapid deceleration of the escape system immediately after separation from the launch vehicle. Additionally, the impact of the opening of the recovery parachute may be quite severe at these higher airspeeds.

Two basic types of emergency escape systems have been used to assure spacecraft crew safety. The impact environments associated with each type are different in many aspects. The first type, the individual ejection seat such as is used in high speed aircraft generates short duration acceleration pulses throughout its entire sequence. These pulses are created during ignition of the ejection catapult, firing of the sustainer rocket, impact with a high velocity airstream, parachute opening shock, and landing impact. The second type of escape technique involves the propulsion of the entire spacecraft away from the launch

vehicle. The catapult acceleration required with this type of escape system is generally of a lower magnitude, usually no greater than 8 to 12 g, and longer duration than that required if an ejection seat is used. The alignment of the propulsion system thrust vector and center of gravity of the vehicle are more easily controlled than is the case for the conventional ejection seat. A large portion of the ejection velocity must be imparted to the ejection seat while it is still stabilized by the ejection rails due to this problem. Therefore, the ejection acceleration may be as high as 18 to 20 g.

Using the entire spacecraft as an escape system causes two other notable differences in the impact environments. The first, a beneficial difference, is the elimination of the problem of impact with the windstream and rapid deceleration. The spacecraft is generally optimally designed for aerodynamic deceleration for reentry into the earth's atmosphere, and thus, the deceleration forces tend to be low. The second difference occurs at landing. Landing without the spacecraft is usually accomplished without incident by a properly trained crewman. A crewman descending under a personnel parachute may judge his drift rate and even control his direction of drift. He can thereby position himself and use his legs to minimize the effects of landing impact. Assuring an equally safe landing of the spacecraft under emergency and even normal conditions is a difficult design problem.

The relatively complex tasks performed by an individual prior to a parachute landing, that is, sensing drift rate and direction and aligning himself to obtain the best use of his legs to attenuate the impact, are tasks that are not easily accomplished without adding

undesirable weight and complexity to the spacecraft. The impact accelerations that are experienced during capsule landing impact are quite variable due to the lack of control of these factors. Furthermore, the variabilities of the spacecraft structural rigidity, the stiffness and contour of the impact surface, and the oscillation induced by the recovery parachute, coupled with the possibility of multiple impacts in different directions add to the difficulty of providing a safe landing.

An escape system composed of several of the most desirable attributes of each of the basic escape system approaches represents another alternative. This approach uses the spacecraft to achieve separation from the launch vehicle but individual ejection seats are also used after the separation distance that is required is achieved and the spacecraft velocity has decayed to an acceptable level. This approach avoids the ground landing impact problems associated with recovery of the crew within the spacecraft; however, it may not be the most effective approach in terms of spacecraft weight and complexity unless there is no requirement to recover the spacecraft.

Impact environments may also be encountered during other portions of the space mission. For example, the acceleration associated with spacecraft docking operations, that is, coupling the spacecraft to another spacecraft or propulsion unit, will result in transient acceleration. The ground landing problem that exists in spacecraft recovery after mission completion or emergency escape also exists during extraterrestrial landing. The impact environments of docking and extraterrestrial landing must necessarily be mild to prevent any

injury to the crew of spacecraft equipment that might compromise the success of the venture.

Each potential or actual impact hazard associated with the mission must be assessed to determine the degree of risk of injury or equipment failure that may be allowed. A mission risk analysis of this type cannot be carried out without a relatively detailed understanding of the human response to each level of impact stress. One of the primary objectives of research in this technology area has, thus, been the development of human exposure limits in terms that are suitable for such a risk analysis.

#### Definition of Impact

Impact is generally defined as an acceleration with a pulse duration of not more than one second. The acceleration-time history is defined in terms of its magnitude in  $m/sec^2$  or usually in g units and time its parameters. The time parameters include rise time (the time duration from start of acceleration to the time of peak acceleration), and pulse duration (total time of individual pulse). Acceleration derivatives such as rate of onset of acceleration (g/sec) and rate of offset of acceleration are also commonly used as descriptors. However, it must be kept in mind that these descriptors give approximations only to the true acceleration-time history and that the limits within which they are meaningful must be examined.

For the purposes of frequency domain analysis, an impact pulse is composed of energy density distributed over a spectrum of frequencies. Thus, a particular acceleration-time history may be reduced to terms of the power spectral density.



Impact accelerations might occur as linear or rotational accelerations, all together in 6 degrees of freedom.

The terminology that is used in the study of the human response to impact is varied (20, 45, 46, 81, 82, 91); however, terminology that is generally understood has been selected for this writing. Terms such as "overload" used in the USSR literature and "dynamic overshoot" used in the literature of the USA are not used to permit a more universal understanding of the text. The direction of the linear and rotational acceleration vectors is defined with respect to the human body by use of the coordinate system shown in figure 2, which is standardized for biomechanics area.

#### Physiological and Pathological Effects of Impact

Most human impact research has been conducted in connection with general automotive or aviation crash research and not in support of specific space requirements. The impact exposures experienced during emergency escape maneuvers have been studied over the last 30 years in connection with emergency escape from aircraft, and the impact situations similar to space capsule landing impacts were of interest for the last 15 years for the development of aircraft capsule escape systems (13).

The primary physiological and pathological effects of impact are caused by localized pressures and the resulting relative displacements of body tissue. The massive stimulation of the entire nervous system in an extremely short time results in various sensations and reactions immediately after impact due to activation of pressure and stretch receptors. These sensations will vary in magnitude depending

on the magnitude of the insult and will vary seriousness from momentary stunning and mild cardiovascular reactions to cardiovascular shock, unconsciousness and concussion, the latter probably always being connected with pathological injury. Direct injuries to the body tissues result when the relative displacements of body tissue exceed the mechanical stress limits of the particular tissue that is involved. These injuries may occur at a cellular or subcellular level with no gross evidence of the shear, tensile, and/or compressive stresses. Damage due to blood movement has not, in general, been observed although conjunctivitis and retinal symptoms observed in  $-G_x$  impact may be related to this phenomenon. This type of injury would not be expected for very short duration impact since the duration of exposure to acceleration is too brief to allow significant shifting of blood volumes. The physical response of the body and its organs, i.e., the stress distribution along the body and stress severity, is dependent upon the acceleration-time history of the impact environment. Other major factors influencing the response include the direction of the acceleration (49, 62), the degree of restraint (64), and the condition of the body, that is, age, physical state, etc. (41, 76). The pathological manifestations described rely heavily on the analysis and interpretation of aviation, automotive, sport, and home accident data as well as data collected from suicides (85); the causes and mechanisms leading to these effects are derived from low-level, noninjurious human tests or animal experiments.

## Head and Neck Injury

Research conducted to date has shown different mechanisms of impact injury and symptoms for each body axis that has been studied. Much of the information that is available on these injury mechanisms has been collected from studies of accidents (71) as well as laboratory experiments. In accidental situations head injury is the most frequent and most severe manifestation (70). More than 75 percent of aircraft crash fatalities result from injuries to the head. These injuries usually occur from heavy blows to the head rather than from the acceleration of the head structure as a whole (2, 3). Injury to the neck, specifically to the cervical spinal cord with concomitant concussion, apparently occurs as a result of hyperflexion or hyperextension of the neck if the head is not supported during impact (19, 40, 66). Other types of concussion are observed after concentrated blows to the head that deform or fracture the skull (31, 32, 56) and cause strains throughout the brain tissue (11, 33, 52, 63).

## Injury from Longitudinal Impact

Damage to the vertebral column is a common mechanism of injury where the impact is applied parallel to the spine in the +G<sub>z</sub> direction as in seat ejection maneuvers (18, 98). Compression fracture of individual vertebral bodies is frequently observed in radiographic examination of individuals who have used aircraft ejection seats (35). These fractures are usually confined to the upper lumbar and lower thoracic areas of the vertebral column. Although such injuries to the upper thoracic and cervical spine are relatively uncommon, they are observed when the ejecting crewman is poorly positioned prior to

ejection. The immediate symptoms of this injury may range from slight pain to severe, incapacitating pain. Illious, persistent neuralgic and sciatic-like pains are common lingering symptoms. Compression fractures or fractures of the spinous processes may, in extreme cases, be sufficiently extensive to result in intrusion of bone fragments or the disc in the spinal cord canal. Such instances may result in paralysis or other neurological symptoms.

The physiological and pathological effects of impact in the  $-G_z$  direction have not been identified in humans (18). Investigators have speculated that intracranial hemorrhage would be the limiting factor on the basis of results of longer duration acceleration experiments. However, impact tests with animal subjects have not supported this theory. Experiments with volunteers have been limited to tests required to support the development of the downward ejection seat and evaluation of Project Apollo crew protection designs (10, 37).

#### Transverse ( $+G_x$ ) Acceleration

When the impact is transverse to the longitudinal axis of the sitting well supported and restrained body, symptoms of various degrees of shock, that is, pallor, perspiration, and transient elevation and subsequent drop of blood pressure, have been the first signs of limiting human tolerance (5, 87, 89). In one test, brief attacks of low blood pressure and albuminuria were observed for about six hours after the impact. More severe impacts will result in unconsciousness. The effects of the maximum voluntarily tolerated impact levels were sometimes not pronounced, but delayed effects occurred with gradual onset over the following 24 hour period. Subtolerance impact exposures

in this axis normally cause an elevation of pulse rate to approximately 150-170 pulses per minute with a respiration rate of 30 to 40 per minute followed by a rapid drop in these rates. Upon repeated exposures the degree of these functional changes before and immediately after impact is decreased (25, 25, 63, 64).

The bradycardia and extrosystole which occur in the first seconds after impact may be indicative of traumatic effects. The disturbance of cardiac rhythm in white rats accompanied, as a rule, damage to internal organs (27, 63). However, bradycardia has been observed immediately after exposures of human subjects to  $-G_x$  and  $+G_x$  impact levels as low as 15 g (96). This response was related to activity of the vagus nerve, since atropine has been shown to block the bradycardia. Test subjects also exhibited transient neurological symptoms for brief periods after exposure to impacts in the 15 to 25 g range in the  $\pm G_x$  direction.

Although physiological stimulation may be hormonal or neural in origin, the immediate onset of bradycardia in response to impact is consistent with neural stimulation. Cardio-inhibitory reflexes of the body can be initiated from baroreceptors in the aortic arch carotid sinus and by visceral afferent nerves originating in nearly all tissues and organs except the skin may produce bradycardia (77). Stretch receptors in the lung can initiate reflex cardiac slowing (18).

Stretching or distortion of the lung tissue can occur during  $-G_z$  impact and may be the cause of the bradycardia observed in tests in this axis. Vascular fluid shifts are an unlikely source of stimulation to the cardio-inhibitory reflex areas because of the brief duration

of impact. However, it is apparent that the inertial effects of  $-G_z$  impact would produce a transient increase in the hydrostatic pressure sensed by the baroreceptors, which in turn respond to this pressure increase by reflex slowing of the heart rate.

Evidence of damage to the respiratory system is also seen in impact studies. The injury ranges from minor functional changes in maximum ventilation of human subjects within voluntary exposure levels (34) to contusion and hemorrhage in animal subjects at near-lethal levels (7). Restraint straps and structures may themselves be responsible for lung damage seen in some of these experiments (7, 23, 78).

#### Lateral ( $\pm G_y$ ) Acceleration

There is a general lack of controlled experimental data on the physiological and pathological effects of lateral ( $\pm G_y$ ) impact. Prior to the emphasis placed on this particular problem by designers of space vehicles, knowledge of the effects of lateral impact had been limited to accident data and data from centrifuge experiments in which long duration acceleration up to 10 g was shown to be tolerable (36). Radiographs collected during these experiments showed extensive displacement of the thoracic and abdominal viscera at acceleration levels as low as 6 g. In support of specific space flight requirements rhesus monkeys were subjected to impacts of up to 75 g at velocities up to 9.8 m/sec with and without contoured lateral support without observing post mortem evidence of injury (73). Electrocardiographic evidence of transient changes in both conduction and rhythm was noted at higher accelerations and impact velocities. Comparison of radiographs

taken before and after impact revealed a displacement of the heart in the direction of the inertial response; however, sequential radiographic observation indicated that the heart returned to a normal position within about 3 hours after impact.

#### Response to Angular Acceleration

Angular impact acceleration may occur during the initial phase of ejection when the escape system separates from the ejection rails or during landing impact of the spacecraft (13, 93, 98). Studies of the physiological and subjective response of volunteer subjects have been limited in the range of environments that have been explored by the capabilities of motion simulation devices that have been used. One study, conducted with acceleration durations of 0.2 to 0.22 second and braking durations of 0.25 to 0.25 sec, explored acceleration levels up to 534 radians/sec<sup>2</sup> with rotation about a "side to side" axis close to the seat-man center of gravity (95). Limiting symptoms were manifested as hyperemia indicating that the limiting factor for the range of acceleration amplitudes and durations explored thus far is the inertial forces within the cardiovascular system acting within the head. Angular accelerations up to 1089 radians/sec<sup>2</sup> with a duration of 0.2 sec (braking deceleration was 816 radians/sec<sup>2</sup> for 0.25 sec) were well tolerated when the rotation was about the longitudinal axis of the body. The effects of angular velocities up to 13.1 radians/sec have been studied with exposure times of several seconds (100). These velocities were tolerated when the axis of rotation was through the heart. When the axis of rotation was through the center of gravity of the man, i.e., through the abdomen at the level of the iliac crest,

symptoms in the head approached subjective tolerance at 8.8 to 9.4 radians/sec. The development of conjunctival petechiae was found to be a reliable measure of the stress imposed on the unsupported peripheral vasculature. The curve for conjunctival petechiae, when the center of rotation was at the iliac crest, varied from 3 seconds at 9.4 radians/sec to 2 minutes at 5.2 radians/sec. With the center of rotation at the heart, petechiae appeared only at velocities of 2.7 to 3.1 radians/sec higher for the same durations.

#### Omnidirectional and Repetitive Impact; Cumulative Effects

The unpredictability of the impact vector and the possibility of repetitive impacts during capsule landing in rough terrain or severe sea conditions necessitated various studies with oblique impact vectors. Although these results are by no means very conclusive or exhaustive, they proved the safety of limited, anticipated impact profiles and precluded the occurrence of unexpected biological effects (62, 90, 99).

So far no evidence of cumulative effects of several successive impact exposures in the same or different directions close to voluntary limits has been reported. However, the number of subjects and exposures are too limited, and physiological and psychological tests are too crude to permit valid differentiation of subtle effects of such stress from the changes which occur with time in individuals unexposed to impact. However, experiments designed to study the pathology associated with repeated impacts have been accomplished with white rats (24, 29). This study was performed with impacts up to approximately 600 g at 1.2 to 0.8 millisecond durations. Accelerations of 450 to 600 g were



applied at 2 to 3 minute intervals in one series of experiments and in 1 and 24 hour intervals in a second series. The animals were impacted from two to 14 times. Impact velocities were varied from 4 to 7 meters per second. The cumulative lesions resulting from repeated exposures at 1 hour intervals were detected as primary lesion of the lungs. Lesions developed after a comparatively small number of repeated exposures.

#### Research Approaches to Studying Tolerance Limits

Numerous approaches have been used in research to determine the physiological effects of impact and to quantify impact exposure limits. Early studies of man's reaction to impact conducted during and immediately after World War II were directed toward answering questions concerning the safety of ejection seat catapults (1, 18, 75, 80). Extensive experimentation was also accomplished to study the effects of aviation crash landings and the short duration  $-G_x$  deceleration encountered during ejection from a high speed aircraft (87, 88, 89). Most of this early experimentation was done with human subjects; often the investigators themselves. Although anthropomorphic dummies were used to evaluate the adequacy of the experimental apparatus before the tests with volunteer subjects, the usefulness of the data collected with dummy subjects was very limited. Animal tests were also performed but the value of these tests was minimal and at best qualitative due to the paucity of information that might be used to relate the relative impact tolerances of man and animals. For the most part, the significant work accomplished at this stage in the development of aviation medicine was based on the subjective comments

of the volunteers, usually mild and often vague symptoms, and the judgments of the investigators conducting the experiments. This approach continues to be used to define voluntary tolerance limits and to evaluate the relative merits of protection systems, but refinements of methodology and a more substantial scientific literature have somewhat reduced the risk associated with this approach.

Impact testing with animals has become a more meaningful approach to assess the effects of specific impact environments and to recognize and analyze specific injury patterns as the volume of data collected with each species has increased. Experiments with animals have been used to provide a basis for estimating the types of injury that might be expected for different acceleration directions and variations in protective equipment configurations (27, 42, 53, 55, 61, 84). Animal tests conducted to determine the frequency of lethal injury have served to substantiate theories of the biomechanical effects of impact, that is, the deformation of load bearing tissues and the effects of the impact-time parameters on the attainment of injurious levels (47, 48). Whereas animal data were originally only of qualitative use in identifying injury patterns and mechanisms, their quantitative usefulness had to wait for the establishment and verification of dimensional scaling laws based on broad progress in the biomechanics area. The validity of these scaling relationships are supported by tests with various types of mechanical stimuli such as airblast, vibration, and sustained acceleration (7, 101, 102).

Despite the advancements that have been made in this aspect of impact research, data collected from animal experiments must be

approached with more than an ordinary degree of caution. Basic differences in anatomical geometry on both a macro as well as a microscopic level undermine the fundamental scaling requirements for similitude of structure geometry and material properties.

Furthermore, not only may the dimensional proportions of the animal be significantly different than man but, perhaps more importantly, the physiological responses may be manifestations of other dissimilarities.

Another approach that has been used to determine impact limits without actually endangering living subjects is the use of human cadavers or tests of tissue or organs taken from cadavers. This approach has been more successful in the study of the breaking strength of bone since the post mortem changes in bone are less pronounced than in soft tissue. Impact exposure limits for the +G<sub>z</sub> direction have been developed at least partly on the basis of tests conducted on cadaver vertebral segments (28, 75, 92). Much of the work that is available on head injury (31, 33, 50, 52, 79) has been obtained from tests conducted with cadaver skulls.

Contemporary biomechanics research has progressively become more directed towards the establishment of impact exposure limits in terms of probabilities of injury and/or fatality instead of the earlier used oversimplified concepts of "limit of tolerance" or "zone of injury". Such relationships can only be obtained by the integration and correlation of all six basic approaches: (1) experimentation at low impact levels using volunteer subjects to establish the basic kinematics of the living body and its relationship to the kinematics of animal and cadaver bodies; (2) discovery of the areas of injury, mechanisms

of injury, and severity of local impact by using cadavers at high impact levels; (3) experimentation with animals to study the full range of physiological and pathological responses in various species; (4) analysis of human accident data to verify laboratory research and to clinically evaluate the severity of the injury and the longer term outcome of these injuries; (5) testing of isolated components of the human body such as vertebral segments or skulls to determine the mechanical properties, i.e., breaking strength, stiffness, etc.; and (6) integration of results from (1) to (5) into a theoretical framework or mathematical model, which allows prediction of response dynamics and injury probability for exposure parameters not yet experimentally tested.

One major difficulty in determining useful impact exposure limit criteria is that the impact levels are not determined by the biological system alone but are strongly influenced by and coupled to the body support or restraint system used in applying the mechanical force to the man. A definition of impact exposure limits without definition and accurate description of this support and restraint is meaningless. The physical dimensions and mechanical properties of all contact areas, that is, seat, backrest, restraints, head support, etc., must be controlled and described with the test data. In the case of animal experiments these "mechanical components" must also be scaled dimensionally, dynamically, and in strength to allow meaningful extrapolation to the human case.

## Mathematical Models

The application of models to represent the dynamic responses of the human body and the support and restraint system can be of great value in determining the relative effects of specific characteristics of the human or his mechanical protection system elements in impact environments (67, 68, 83). Furthermore, their use enables one to analytically determine the effects of complex waveforms that could not otherwise be described by such simple parameters as peak acceleration, risetime or rate of onset that may be used only as descriptors of relatively simple waveforms.

The various models that have been developed have had one of the following purposes: (1) to understand the basic pathological, physiological or anatomical dysfunctions resulting from impact; (2) to extrapolate from environments evaluated in the laboratory to operational environments not yet tested; (3) to determine optimum protection system designs for a given set of environmental parameters; and (4) to use the model to evaluate and interpret tests on human surrogates, i.e., animals or anthropomorphic dummies. The general types of biodynamic models may be categorized as models that describe the properties of tissue, human body subsystems such as the head and neck, the total body response, or the kinematic response of the whole body. Models developed to describe experimentally obtained tissue properties provide a basic understanding of the basic physical processes by which mechanical energy is transmitted through the body tissue in various frequency ranges (21, 44). Subsystem models of the human body such as mathematical representations of the head (11) and spinal column (68, 28) have been

shown to have the greatest degree of practical usefulness. Models of this type have been used to account for the statistical variability of failure modes and the effects of parameters such as the age of the individual (92). The total body model is composed of several of the subsystem models and allows a more complete understanding of the interaction of the various responses. The model shown in figure 3 is an example of such a total body model developed to combine the body's response characteristics in the  $G_z$  direction as measured in both vibration and impact exposures. For a more detailed analysis of specific injury modes, it is often preferable to use subsystem models where further refinements and nonlinearities can be investigated more easily. The kinematic models that have been used depict the individual segments of the body as a linkage system with individual components having the geometric shape and inertial properties of the human body segments and the degree of joint mobility as well as muscle forces that have been derived from experimentation (58). Such models are useful in determining the motion of the body segments of crewman during specific impact conditions and in predicting the interaction of the body segments with the restraint system and the interior surfaces of the spacecraft.

The interpretation and application of the relatively large amount of available data on the effects of impact on subhuman primates and other mammals is vitally dependent upon the use of model scaling techniques. The basic assumption of this approach is that an impact environment will lead to similar injury mechanisms in animal and man when dynamic similarity or scaling laws are applied. This assumption

must be continually verified when attempts are made to use this approach in light of the geometric dissimilarities between species. Methods commonly employed in such verification include evaluation of the similarity of the mechanical properties of tissue, steady state vibration response analysis of various species of different size, kinematic response to impact, and evaluation of the injury mechanisms observed in clinical investigations of humans involved in accidents where the impact environment can be reasonably estimated. Although the anatomical and physiological differences between the various species and the assumption of similarity of injury mechanisms may present sizable obstacles, valuable first approximation results can be obtained from using scaling laws. Applying the scaling laws given in figure 4 one may obtain the approximate resonant frequencies for the chest, spinal, and abdominal systems for various animal species as shown in figure 5.

#### Impact Simulation Techniques

A wide variety of mechanical facilities have been used to simulate the impact environments anticipated in normal and emergency space flight operations. To assure broad usage of the test data and their mathematical interpretation and easy application to biodynamic models most work has not been conducted with the complex acceleration waveforms encountered in actual operational situations but with simple approximations to these patterns such as rectangular, triangular, and half sine pulses. The simplest of the facilities used are the vertical deceleration towers. These devices use gravity to assure the reproducibility of the impact velocity. The impact-time history may

be controlled by using hydraulic decelerators (12, 99), crushable materials such as aluminum honeycomb, or energy storage devices such as elastomeric materials or liquid springs.

Ejection towers have been used since immediately after World War II to study man's response to  $+6g$  acceleration, evaluate personnel protective equipment and to provide crew training (1, 18). These towers have incorporated both pyrotechnic and pneumatic devices to accelerate the ejection seats and subjects. Rocket powered sleds, propelled along horizontal tracks into water brakes, have been used to study the combined effects of short duration deceleration and windblast encountered during emergency escape from high speed aircraft (1, 87, 89). More precise studies have been accomplished using a pneumatically propelled sled and water brake decelerator that was designed for the purpose of conducting human tests (12). This facility is shown in figure 6.

Other impact simulators include simple pendulums and pneumatically powered strikers. The pendulum impact devices have been used to study impact protection systems (51) and to study head impact tolerance as well as evaluate protective headgear. Special small scale pneumatic strikers have been developed to study head and thoracic trauma (66).

#### Safety Precautions During Experimentation

The impact simulators must be designed to provide precise control of the impact environment parameters if human subjects are to be used at impact levels approaching tolerance. Reproducibility of the test environment is especially critical in experimentation where the impact stress is increased in small increments until voluntary tolerance is



reached. Furthermore, the test apparatus used with the simulator must receive extraordinary care in its design and the understanding of its contribution to the test results. Where prototype hardware, such as an astronaut ejection seat, is used one must recognize that the design of the structure of the seat may include only a small margin of safety, say a factor of 1.25, since the impact environment under study would be encountered only under emergency conditions. This margin of safety, while suitable for a low probability of occurrence situation such as emergency escape, is normally not considered adequate for experimentation with volunteer subjects. The rigidity of the structure, or lack of it, is not only important in considering the safety of the apparatus but also in the fidelity with which it transmits the impact of the simulator to the subject. Unfortunately, the acceleration transmission characteristics of the apparatus and component articles such as seat cushions and padding are often ignored. In these cases it is usually difficult if not impossible to draw any general conclusions about the work or to extrapolate to other equipment configurations. Where the determination of human tolerance is the primary objective of the experimentation it is often simpler to assure that the structure is rigid and eliminate elastic padding. Furthermore, the rigid structure lends itself to the repetitious use common for impact testing.

Beyond the more straightforward considerations of experimental procedures and apparatus design lies the fundamental ethical questions surrounding impact experimentation. Perhaps the most basic question: "Is the value of the information resulting from the test commensurate

with the risk to the subject?", should be answered not only in the initial planning stages of the research program but also immediately before the initiation of testing when the scope and adequacy of the data to be collected are more completely defined. In any case the investigators are ethically bound to minimize the risk to the subject. Actions that can be taken to achieve this end include accomplishment of thorough physical examinations prior to and after testing and careful medical-monitoring throughout the experimentation and post test period as well as meticulous attention to operation of the impact simulation equipment and emergency procedures. Post test examination and follow-up of the subjects depends on the specific test goals, subject symptoms reported, and the medical investigator's report.

#### Human Impact Tolerance as Related to Space Missions

During early work on manned spacecraft designs there was recognition of the necessity to acquire more complete data on the human response to impact. The available literature reflected that the majority of the impact research had been directed toward the solution of aviation problems. First, the acceleration exposure limits for the Z axis had been developed as design criteria for ejection seat catapults and therefore were defined in terms of the acceleration waveforms that are normally obtained from such ballistic devices. Second, the X axis limits were similarly defined for pulse shapes that were anticipated during the deceleration of ejection seats immediately after ejection into high velocity windstreams. Third, practically no data were available to assess the effects of impact vectors acting in the Y axis. Furthermore, the information that was

available pertained only to the cardinal axes and thus, the effects of impact vectors acting in directions other than these axes could not be evaluated.

The work of Eiband summarized the data that was available within the United States at this time (18). These data are graphically summarized in figures 7 and 8. While these data have been of inestimable value in the development of design criteria for manned spacecraft, they were inadequate for the evaluation of specific impact problems associated with both normal and emergency astronautic operations. As mentioned previously, providing escape from the launch pad with an ejection seat requires the use of a high magnitude, short rise time acceleration pulse. Additionally, and most important, the landing impact environments anticipated during the recovery phase of space missions presented a set of potentially severe conditions characterized by high magnitude, short rise time impact pulses of varying direction and irregular waveform. Unfortunately, the impact exposure environments are hard to predict as long as the prototype space system is not available for test and is always subject to large statistical fluctuations depending on details of the landing conditions. The tolerance limits presented in figures 7 and 8 are only available in terms of idealized trapezoidal waveforms. The deduction of a plateau level and time duration from a complex acceleration-time history encountered in actual practice is not an easy task and in some instances it is impossible.

### +G<sub>z</sub> Impact Exposure Limits

Evaluation of the Eiband summaries shows that there is a considerable unknown region between the area of voluntary human tolerance and the areas of injury. In the +G<sub>z</sub> direction, figure 7a, the unknown area that is shown covers over 20 g in the ordinate and does not show human exposures for time durations less than 0.04 second. Unfortunately, this unknown region includes the impact environments of most interest in space operations. In addition, it is clear that the boundaries are not well defined and a few more data points might change the shape of the curves. Although the data that are plotted are too limited in numbers of tests and control of variables to provide a basis for accurate interpretation (65), the general form of the curve shown in figure 7a merits some comment to provide insight into the general form of the tolerance curve in the short duration region. Note that for impact plateau durations up to 0.007 second the data points dividing the areas of severe injury and moderate injury decrease in a nearly linear fashion on the log-log scale as the time duration increases. The relationship of these data points is as it should be if viewed in terms of the dynamic response of a mechanical system. The use of a mechanical analog seems appropriate in this case since the mechanism of injury that is operationally important is mechanical in nature, that is, compression fracture within the vertebral column.

The simplest analog that has been developed for the study of impact applied parallel to the vertebral column (+G<sub>z</sub>) is a mechanical model composed of a mass, a spring, and a viscous damper (92). The mechanical elements are lumped-parameter elements, e.g., all the mass

of the human body that acts upon the vertebrae to cause deformation is represented by the mass element. A diagram of the model is shown in figure 9. This model is used to predict the maximum deflection and associated force within the vertebral column for any given impact environment. Compression fracture occurs when the force in the spring exceeds the breaking strength of the spring. The properties of the model elements have been derived from existing data. The spring stiffness and breaking strength have been determined from cadaver vertebral segments, and the damping ratio has been calculated from measurements of mechanical impedance during vibration tests (16). The response of the model can be determined for any given acceleration-time history by solution of a second order, differential equation containing terms representing the positions of the mechanical elements with respect to time.

Use of the mechanical model also provides a basis for a probabilistic approach to injury prediction. Since the model reduces the effect of the impact environment to a single parameter, that is, the peak deflection or force in the spring element, a correlation can be determined between this parameter and injury. For example, the breaking strength of vertebral is variable but it can be statistically described in terms of probability of failure (92). This same approach has been used to provide estimates of the relationship between age and breaking strength (69, 92).

An analytical effort has been conducted to determine the degree of correlation between the spinal injury model and the injuries that have been experienced in operational aircraft ejection seats (9). The

relationship between the operational acceleration environments and the actual spinal injury rates of the ejection systems included in the study are shown in figure 9. The response of the model is expressed in terms of dynamic response index (DRI) values. The initial estimate of the probability of injury as determined from cadaver data is compared to the operational data. The slope of the line drawn through the operational data points was established on the variance of the vertebral strength used to establish the initial estimate. The spinal injury model and this injury probability estimate have been used to assess the risk of spinal injury associated with the Project Apollo mission impact environments.

The vertebral failure process has best been described by a mechanical deformation and effect sequence as shown in Table I.

More extensive studies of vertebral and intervertebral disc strength have been conducted to determine more precise estimates of +G<sub>z</sub> impact tolerance (28). This work significantly increases the number of data points since a total of 530 vertebrae were studied. Furthermore, the study included tests of cervical vertebral segments. Heretofore, only a few data points were available to provide an estimate of the breaking strength of the cervical spine. The mean ultimate strength of the vertebral segments tested in this study are given in Table II. The values indicate the same general change of breaking strength as a function of the position of the vertebral segment as similar collections, but the breaking strength is approximately 18 percent higher (69). The data were obtained from vertebral specimens from men ranging in age from 19 to 40 years; less

Table I. Mechanical Failure Sequence of the Vertebral Body Under Axial (+G<sub>z</sub>) Compression (28)

Deformation	Effect
6-10%	Within elastic range of deformation. No macroscopic structural changes.
12-13%	First macroscopic irreversible changes. Compression of limbic zone.
17-18%	Cracks and compression in area of wrist of vertebral body.
25-26%	Fractures within vertebral bodies without displacement of hips.
36-37%	Fractures with dislocation.

Table II. Ultimate Strength of Vertebrae Compressed Vertically (28)

Vertebra Segment	Strength in kg	Vertebra Segment	Strength in kg	Vertebra Segment	Strength in kg
C1	800	T2	436	T10	860
C2	510	T3	467	T11	917
C3	404	T4	522	T12	1054
C4	408	T5	551	L1	1059
C5	453	T6	619	L2	1175
C6	563	T7	681	L3	1269
C7	464	T8	824	L4	1296
T1	475	T9	840	L5	1286

than 30 hours elapsed after death before the start of the experiment. The data that are shown were obtained using a deformation rate of 10 mm/min. The number of observations used to compute the arithmetic means was from 6 to 16.

Average mechanical characteristics of the intervertebral discs of the cervical, thoracic, and lumbar sections of the vertebral column are given in Table III. The ultimate strength was identified by rupture of the fibrous ring of the disc and extrusion of a jelly-like substance.

In connection with longer space missions the potential effects on impact tolerance of prolonged immobilization, physical inactivity, and weightlessness have been of much interest and speculation (30, 76). The cardiovascular and metabolic effects of simulated weightlessness and weightlessness are treated in a separate chapter. It shall only be mentioned here that the cardiovascular changes observed must have some effect on the cardiovascular impact responses described. Quantitative data on this subject are not available and these symptoms are usually not the one considered limiting human tolerance. However, the decrease in bone strength due to osteoporosis of disuse is an established fact and bone loss has been measured on astronauts after space missions and in simulated weightlessness studies on man and animals (57). Although bone loss per se cannot yet be related directly to bone strength there is good reason to assume a noticeable reduction in bone strength after prolonged space missions. In rhesus monkeys immobilized by plaster of paris casts for 240 days a reduction of overall spinal impact tolerance by 25 percent was observed, the



**Table III. Mechanical Characteristics of Intervertebral Discs  
Compressed Vertically (28)**

<b>Vertebra Section</b>	<b>Ultimate Strength</b>	<b>Elastic Deformation</b>
<b>Cervical</b>	<b>486 kg</b>	<b>1.2 mm</b>
<b>Thoracic</b>	<b>1270 kg</b>	<b>1.6 mm</b>
<b>Lumbar</b>	<b>1502 kg</b>	<b>2.1 mm</b>

main decrease in strength having occurred already after 60 days immobilization as shown in figure 10 (43). These data cannot yet be applied quantitatively to an estimate of the strength reduction in human subjects, but it is obvious that they call for further studies and for conservative application of all bone strength/bone impact limits data obtained on "normal" human subjects adapted to the earth's gravitational field.

Tolerance to  $+G_z$  impact applied to the standing subject has been studied to determine the effects of explosions beneath the floor of a vehicle (38). In the case of impact on the sole of the foot with leg extended, fracture of the distal tibia of the human leg was determined to occur at a load of 680 kg applied in axial compression between the knee and foot (39). The limiting velocity change for impact transmitted to a stiff-legged subject is 10 Hertz. The resulting impact exposure limit curve is shown in figure 11. A few empirical studies on cadaver legs are plotted in this figure. Such exposure criteria are of value in the design of lunar or planetary landing vehicles where the crew may be standing upright during landing.

After the initial compressive phase of the response to impact motion of the floor, the unrestrained man will be thrown and will be propelled off the floor with some velocity. While this velocity will not cause injury, it will have a bearing on his velocity at the termination of his motion when injury can occur. The kickoff velocities of men in the standing and seat positions have been measured for a variety of impact pulses (38). The ratio of peak deck velocity,

$V_d$ , to kickoff velocity,  $V_k$ , has been plotted as a function of the ratio of rise time to peak velocity ( $t_p$ ) to natural period of man ( $T$ ) in figure 12. The curves follow the form:

$$\frac{V_k}{V_d} = 2.7 \left( \frac{t_p}{T} \right)^{0.44}$$

Where  $T$  is 0.1 second for the standing man and 0.167 second for the seated man.

#### Transverse Impact Exposure Limits

The effect of impact in the  $-G_x$  direction is critically dependent upon the type of restraint and the posture of the body at the time of impact. Volunteers have been exposed to impact levels up to approximately 45 g for 0.09 second duration with a rate of onset of 500 g/sec (86). The subjects were restrained by 3 inch wide shoulder straps, lap belt, and thigh straps and the subjects' head and neck were preflexed prior to impact. Rate of onset or rise time was found to be instrumental in the production of shock symptoms. Under operational conditions where only 2 inch wide shoulder straps and lap belt are used and the crewman wears a helmet weighing from 1.5 to 2 kilograms, moderate injury may be expected as low as 30 g. In the case of an open ejection seat even higher acceleration levels can be tolerated due to the counteracting effects of aerodynamic drag forces. If the crewman is protected only by a lap belt, impact tolerance is reduced further. Volunteers have tolerated  $-G_z$  impacts up to 26 g for 0.002 second with a rate of onset of 850 g/sec (51).

In the transverse,  $+G_x$  impact direction human tolerance is potentially higher than any other axis if the crewman is restrained

to a full body support. Impact levels up to 36 g for 0.16 second with a rate of onset of 1,000 g/sec have been tolerated by a volunteer subject (85). Severe shock has been observed as a result of a volunteer test at 40.4 g for 0.040 second duration with a velocity change of 14.8 meters/sec and a rate of onset of 2,140 g/sec (4). In view of the available data, the impact exposure limit for the  $+G_x$  direction has been estimated to be 35 g for acceleration durations up to 0.1 second to prevent injury (18). Higher accelerations have been estimated to be tolerable if moderate injury is acceptable.

#### Lateral ( $+G$ ) Impact Tolerance

Human tolerance to lateral ( $+G$ ) impact environments is not well defined. Tests that have been accomplished have explored a rather narrow range of acceleration pulse durations. Volunteer subjects supported by a fully contoured couch have been exposed to impacts of up to 22 g with a rate of onset of 1,350 g/sec where the impact velocity was 19.3 ft/sec (14). Another series of tests were accomplished with volunteers supported laterally by flat plates on which their shoulders would bear during impact (8). The acceleration-time patterns used in these studies are discussed in more detail in the following section dealing with off-axis tolerance.

Tests with volunteers have been conducted with more conventional restraints and seats, but the acceleration levels that were found tolerable were more moderate. A lap belt, shoulder harness and crotch strap configuration was tested with human subjects up to 17.7 g without irreversible injury (72). Tests have been accomplished with volunteers restrained only by a lap belt (103). These tests were

terminated when an acceleration level of 9 g was reached due to prolonged symptoms of pain in the musculature of the neck.

#### Off-Axis Impact Tolerance

Research to define impact exposure limits has been concentrated on the cardinal axes and, therefore, limits have not been developed for impact environments that occur in other axes. Data that are available have been collected to evaluate the acceptability of a somewhat narrow range of impact environments using body support and restraint systems proposed or developed for specific aerospace systems. Table IV summarizes data collected during deceleration tower experiments with seven acceleration vector directions and six acceleration-time histories (8, 99). In these experiments 20 volunteers were exposed to the impact profiles shown in figure 13a. Peak accelerations ranged from 13.4 to 26.6 g with onsets from 426 to 1,770 g/sec. The power spectral density of each of these impact patterns is shown in b of figure 13 and the acceleration directions are identified in terms of the vector orientation numbers designated in figure 14. No injuries were produced in this study although some transient changes were seen in the electrocardiograms.

An additional series of 288 tests were conducted on a horizontal decelerator to supplement the above study and to evaluate the 17 other impact vector directions shown in figure 14 (10, 90). Two tests were performed at each position and g level. The acceleration magnitude and rate of onset increased simultaneously. Accelerations measured on the impact sled ranged from 5.5 to 30.7 g, the rate of onset varied from 300 to 25,000 g/sec, and the velocity of the sled

Table IV. (To be provided.)

at the time of impact ranged from 2.8 to 13.7 meters/sec. A restraint system proposed for Project Apollo was used for these experiments as well as the experiments described in the paragraph above. Table V summarizes the significant findings of the post impact physical examinations of these tests. Volunteer test data are also available for a series of 11 impact tests in a less restrictive body support and restraint system used in Project Apollo (74). During the studies of Apollo restraint and impact vector directions several cases of trauma occurred (90). A forward facing subject tipped back at 45 degrees (position number 5) sustained simultaneous compression and hyperflexion of the trunk which produced persistent soft tissue injury in the area of the 6th, 7th and 8th thoracic vertebrae. The impact was 25 g at 960 g/sec in 0.097 second.

#### Impact by Missiles

Injuries due to the impact of objects propelled by blast pressures, winds ground or floor shock, etc. are dependent upon a number of factors. Among them are the mass, velocity, character, density, and angle of impact of the projectile whether or not penetration occurs; the area and organ of the body involved; the amount and kind of clothing, and the immunological status and general health condition of the injured individual (15, 101). Studies of tissue damage by impact of small objects show that the energy of small objects striking a body surface overlying soft tissue is absorbed in the surrounding tissue and does not bring about motion of the whole body (11, 22)

Tentative criteria for missile damage in humans are shown in Table VI.

Table V. (To be provided.)



Table VI. Tentative Criteria for Indirect Blast Effects Involving Impact From Secondary Missiles

<u>Kind of Missile</u>	<u>Critical Organ or Event</u>	<u>Related Impact Velocity m/sec</u>
Nonpenetrating		
4.54 kg object	Cerebral Concussion:*	
	Mostly "safe"	3.05
	Threshold	4.58
	Skull Fracture:*	
	Mostly "safe"	3.05
	Threshold	4.58
	Near 100%	7.02
Penetrating		
10-gm glass fragments	Skin Laceration:**	
	Threshold	15.3
	Serious Wounds:**	
	Threshold	30.5
	50%	54.9
	Near 100%	91.5

\*References 52, 33, 104

\*\*Represent impact velocities with unclothed skin. A serious wound arbitrarily defined as a laceration of the skin with missile penetration into the tissues of depth of 10 mm or more.

## Impact Protection

The impact tolerance of man or animal is dependent upon the manner in which the impact stress is transmitted to the body and the degree of body support and restraint that has been provided. The method of fixation of the subject to the impacted structure is perhaps the most fundamental consideration. The seat structure and restraint reinforce the body to prevent injurious hyperflexion or hyperextension of anatomical joints and excursions of body organs (59). The body support and restraint acts to distribute the impact loads over the body surface. These loads should generally be distributed uniformly over as wide an area as possible to avoid concentration of pressure. An exception to this rule would be the case where the body may act to attenuate the load being transmitted to vital parts whereas direct coupling might be more injurious. As mentioned previously the method of body support and restraint is a fundamental factor in determining the probability of injury for any given impact environment.

Experiments with mice and dogs immersed in water and congealing gypsum have shown that impact tolerance may be increased, up to 6 times higher than without immersion (59, 60, 61). Covering the walls of the immersion vessel with porous rubber was a critical factor in animal survival (60).

Lap belt, shoulder straps, thigh straps, hand holds, toe straps, and seats of varying degrees of contouring have been used in spacecraft applications to provide crew protection. Many of the early studies of impact tolerance for space flight operations used molded couches of rigid plastic foam to support both animal and human

subjects (64, 73). More contemporary efforts have used simpler seat structures to enhance the interchangeability of crew stations throughout long duration flights (74). These seat structures are supported within the spacecraft by impact attenuating struts, <sup>as shown in figure 15a and b</sup> A variety of attenuation devices have been studied ranging from simple, crushable honeycomb structures to more complex hydraulically damped spring systems and cyclic strain mechanisms (54, 94).

Impact protection can be accomplished to the greatest extent within the impact transmission pathway to the crewman with devices such as impact attenuating struts. Crushing of the vehicle structure provides some energy absorption and this characteristic can be enhanced by the vehicle structure designer. The deformation of cushioning materials and the restraint system can also be designed to minimize the transmittal of energy at frequencies where the human is most sensitive; however, care must be taken to assure that these elements of the protective system do not in fact amplify the accelerations transmitted to the body.

Mathematical models of both the protection system as well as the human body have greatly improved the designer's capability to select appropriate materials for crew seat cushioning and restraint systems (67, 68). These same modeling techniques have provided insight into the effects of the initial conditions of the crewman within his personal equipment. For example, these analytical techniques have demonstrated the importance of eliminating slack or deadspace between the crewman and his body support and restraint system, ~~as shown in figure 15a and b~~

~~Figure 15a~~ Similarly, these techniques have proved design criteria for restraint harness tensioning devices, ~~as shown in Figure 15b~~.

Other methods of crew protection which have been considered include crew conditioning and the use of pharmacological agents. Conditioning of the crew has been considered from several aspects. First, by assuring the best physical condition of the crewman by a sound program of physical exercise before the flight. Second, by crew training and exposure to mechanical stresses during simulated missions. And finally, where long duration missions may cause deconditioning of the musculoskeletal system, exercise and in the future, perhaps, the use of chemotherapy may be used to retard the deconditioning.

#### Summary

The degree to which impact accelerations are an important factor in space flight environments depends primarily upon the technology of capsule landing deceleration and the weight permissible for the associated hardware, that is, parachutes or deceleration rockets, inflatable air bags or other impact attenuation systems. Safe capsule landings on any type of terrestrial and extraterrestrial surface must be the goal of these hardware developments so that the restrictions imposed in this respect on most of the past USSR and USA space missions can be relaxed. However, design for emergency situations such as crew escape during unforeseen failure on the launch pad will always require the most accurate information available on the limits of human tolerance and the risk involved.

Although a considerable body of information has been available on human tolerance to impact and impact protection from aircraft escape, and aviation as well as automotive crash research, the USSR as well as USA space programs had to define specific limits of human tolerance with higher accuracy and reliability than they were known before. Particular contributions in this area include: (a) exploration of impact tolerance for all impact directions, (b) definition of probability of injury for low probabilities of injury consistent with the high reliability/safety requirements of space missions, and (c) development of mathematical models to predict injury probability for complex acceleration functions and to calculate the crewman's biodynamic response when coupled to various support and restraint systems. These advances as well as experiences with new impact attenuating crushable materials and structures are of significance beyond the specific realm of space biotechnology.

The problem most specific to space medicine is the potential change of impact tolerance due to reduced bone mass and muscle strength caused by prolonged weightlessness and physical inactivity. Although valuable contributions to this area have been made through animal experimentation in the USSR and the USA, it requires considerably more research as space missions will be extended over many weeks and months. The relationship between bone strength, bone mass and muscle strength must be explored as a function of gravitational load, isotonic and isometric exercise, time pattern and diet; for osteoporosis of disuse appropriate time scaling factors for bone dynamics as a function of gravitational exposure and activity time

patterns must be established relating animal experiments to human conditions. Changes in injury patterns due to these changes in the musculoskeletal system must be known and understood. Based on such studies the proper impact limit values, protection equipment, preventive measures, such as exercise and possibly chemotherapy, and post flight care can be selected.

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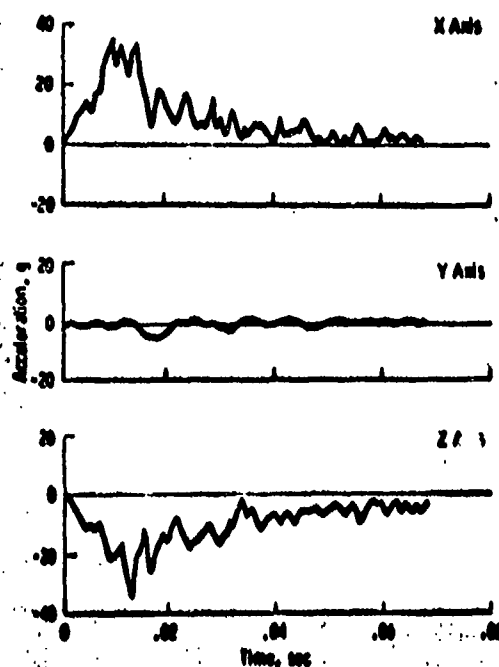


Figure 1. Accelerations Recorded During Impact Tests of the Apollo Crew Module

Impact occurred at a pitch attitude of  $-27.5^\circ$ , a roll attitude of  $0^\circ$ , a horizontal velocity of 11.4 m/sec, and a vertical velocity of 10.5 m/sec.

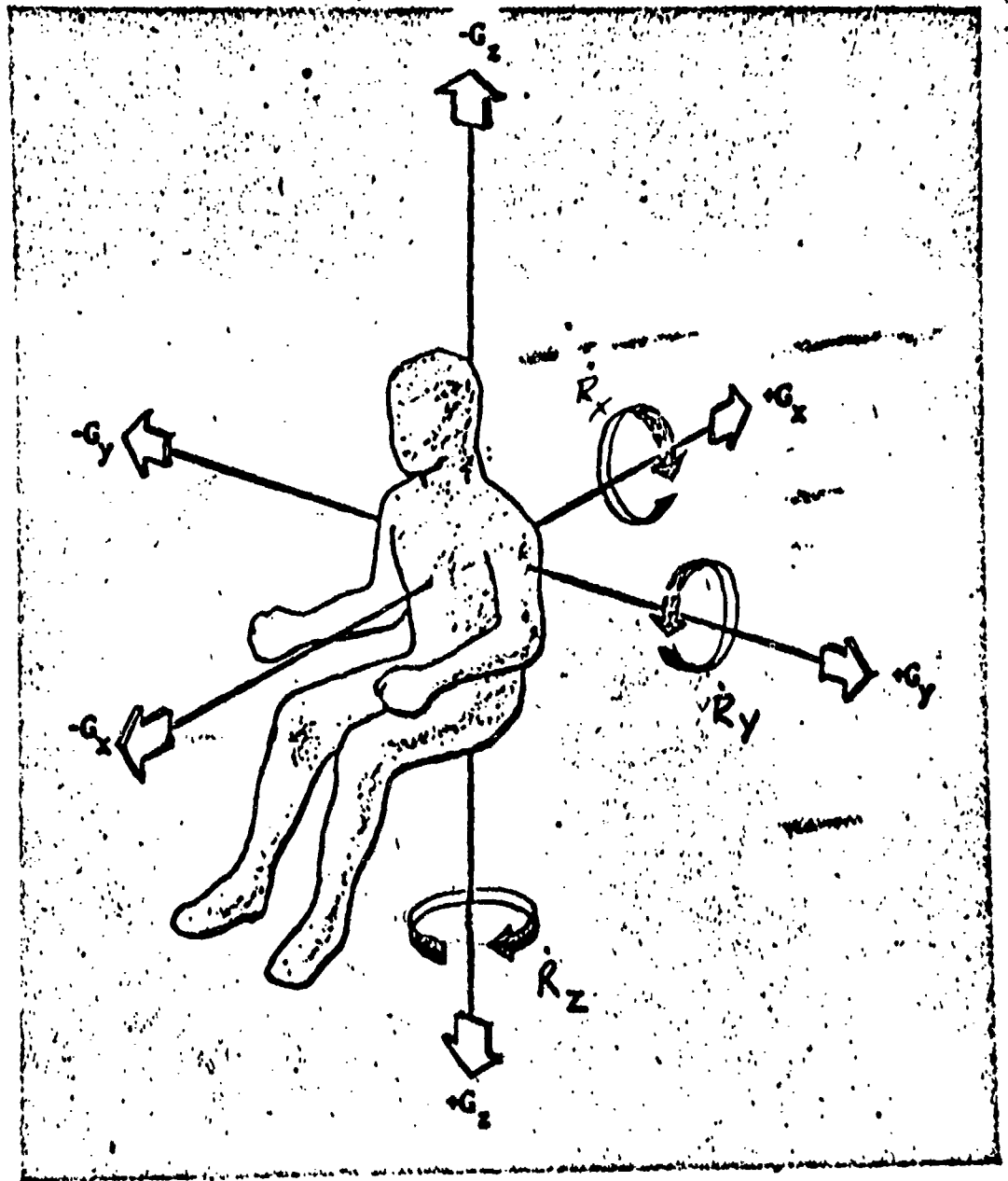


Figure 2. Coordinate System for Description of the Impact Inertial Response Vector Direction

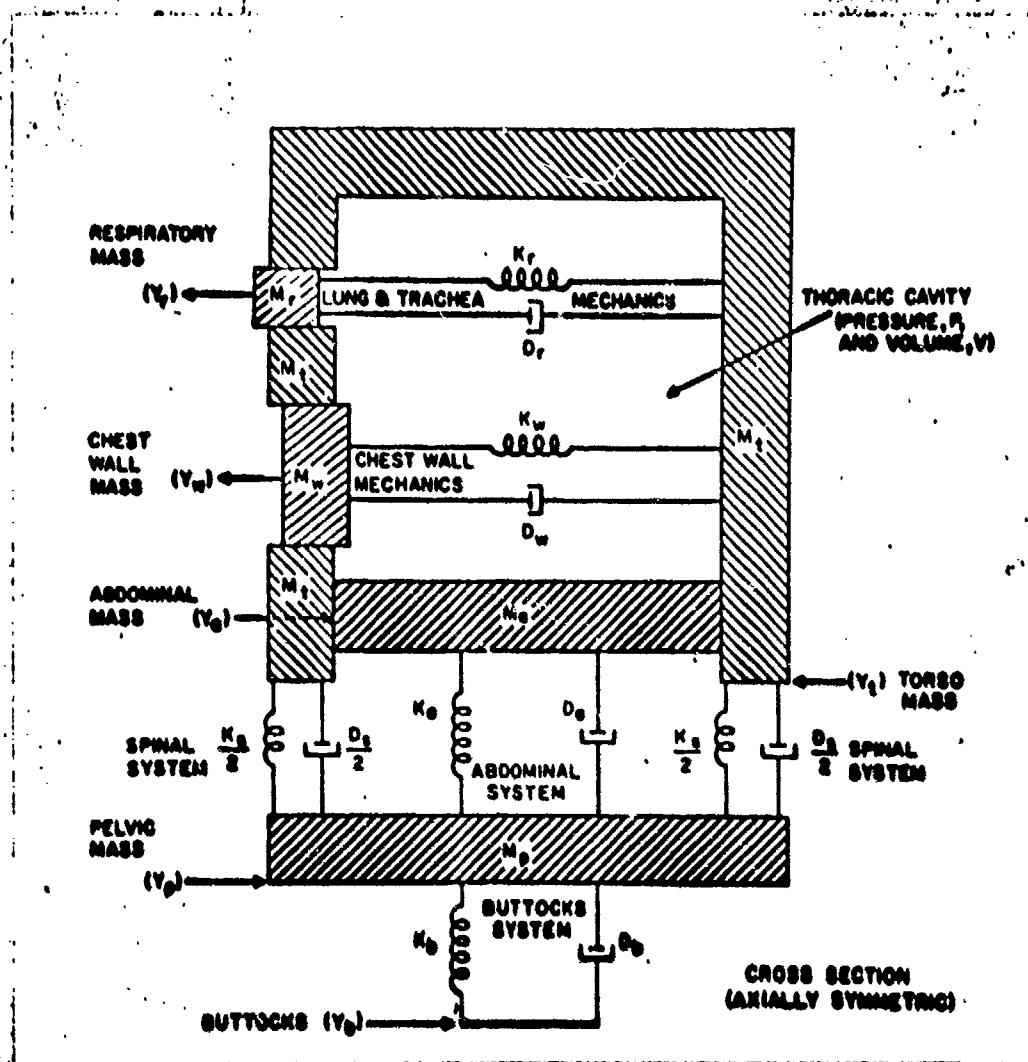
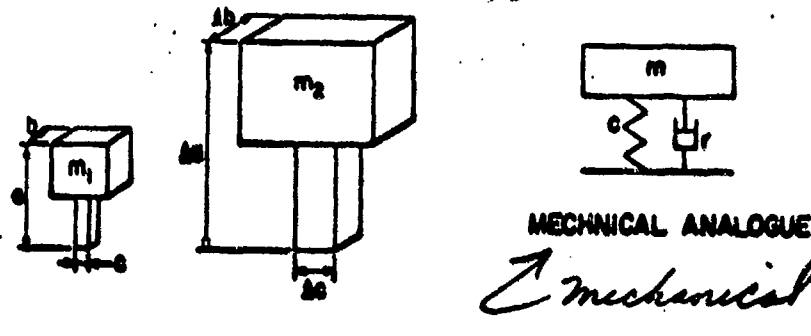


Figure 3. Multidegree of Freedom Model to Depict Whole Body Response to Impact



	TISSUE	AIRSPACE
$\frac{m}{L^3}$	$\frac{m}{L^3}$	$\frac{m}{L^3}$
$\frac{c}{L^2}$	$\frac{c}{L^2}$	$\frac{c}{L^2}$
$\frac{r}{L}$	$\frac{r}{L}$	$\frac{r}{L}$
$f$	$f$	$f$
DAMPING $\frac{1}{L^2}$	$\frac{1}{L^2}$	$\frac{1}{L^2}$
		CONSTANT

Figure 4. Scaling Laws for Geometrically Similar Structures  
Such as Mammals of Different Size

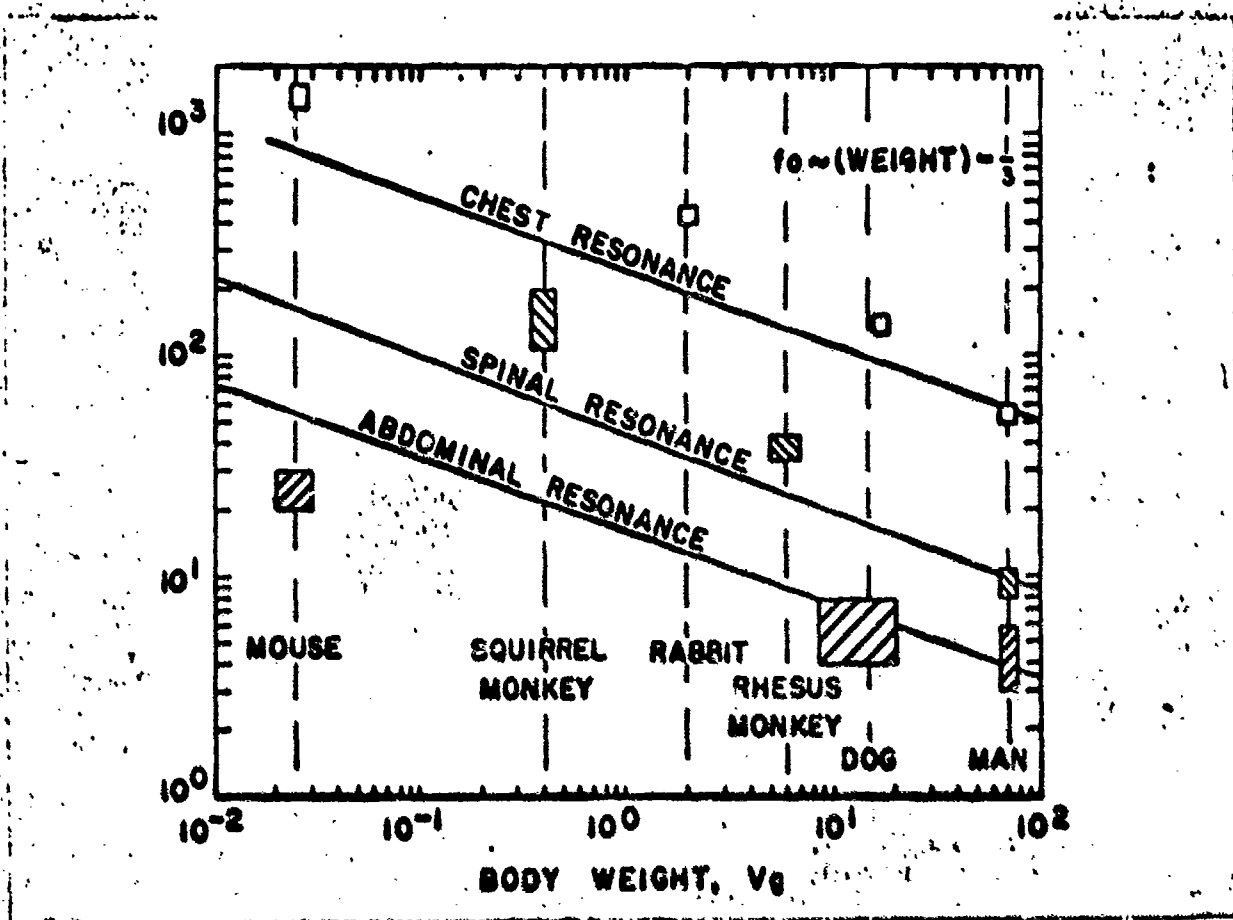
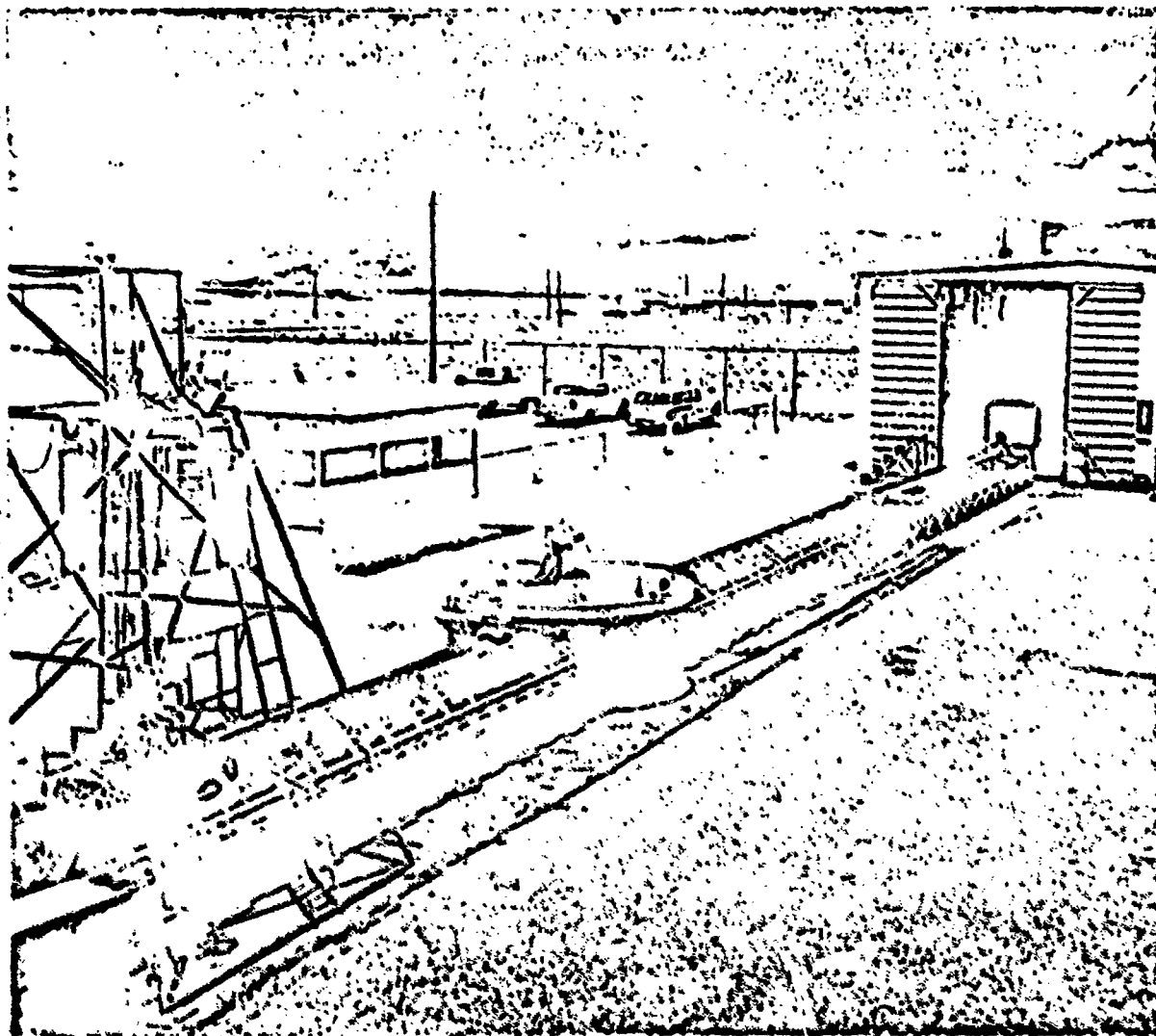


Figure 5. Approximate Resonance Frequencies of Total Body Response Models as a Function of Body Size (Weight)

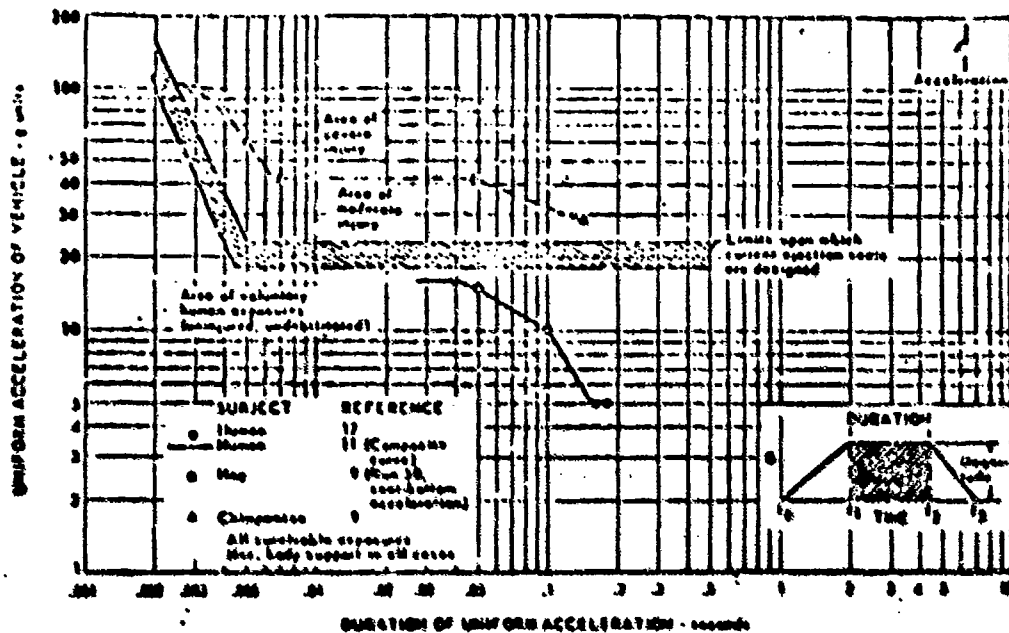


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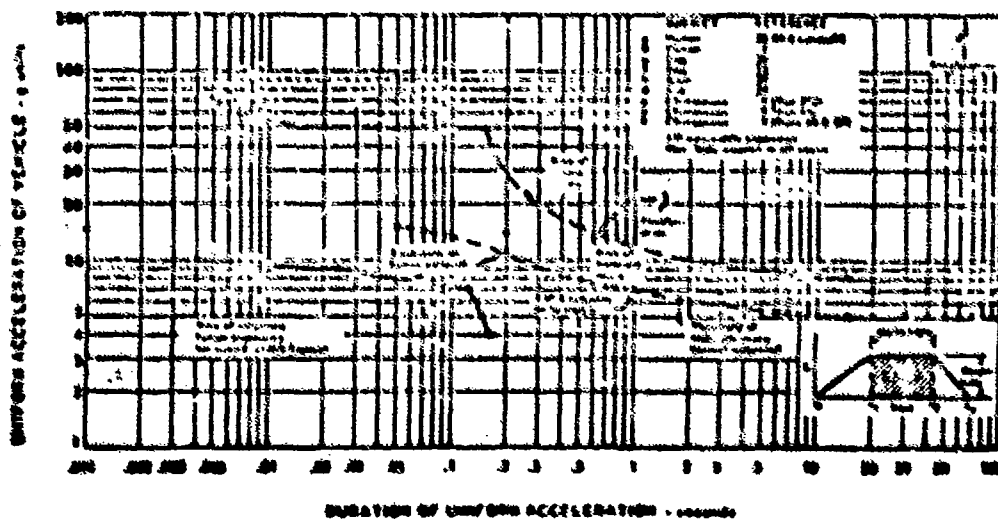
Figure 6. The Daisy Decelerator



7 a.

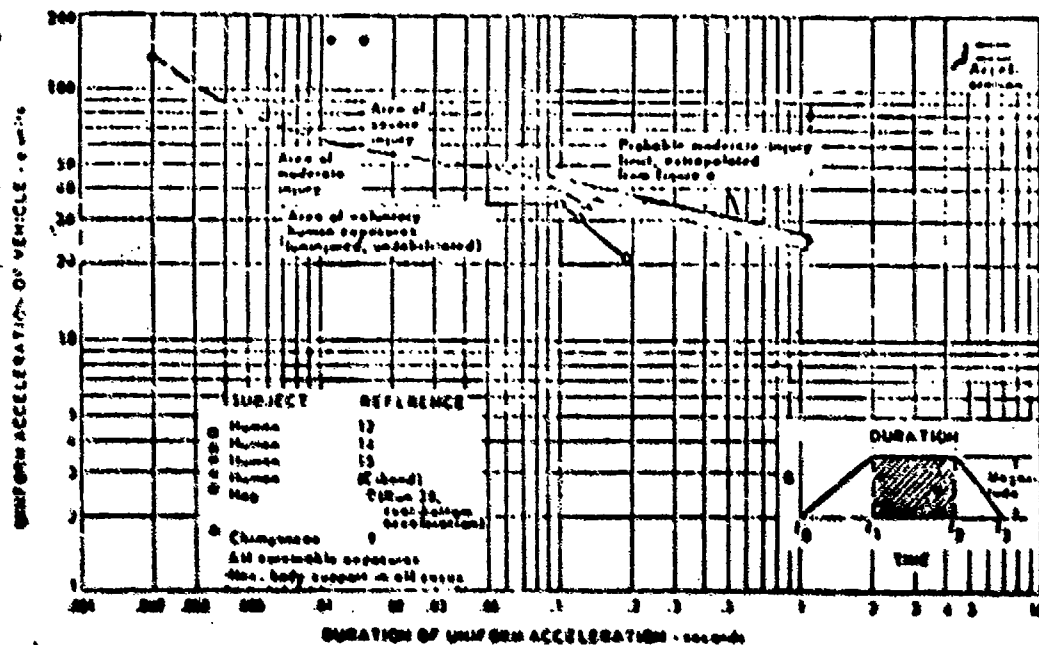


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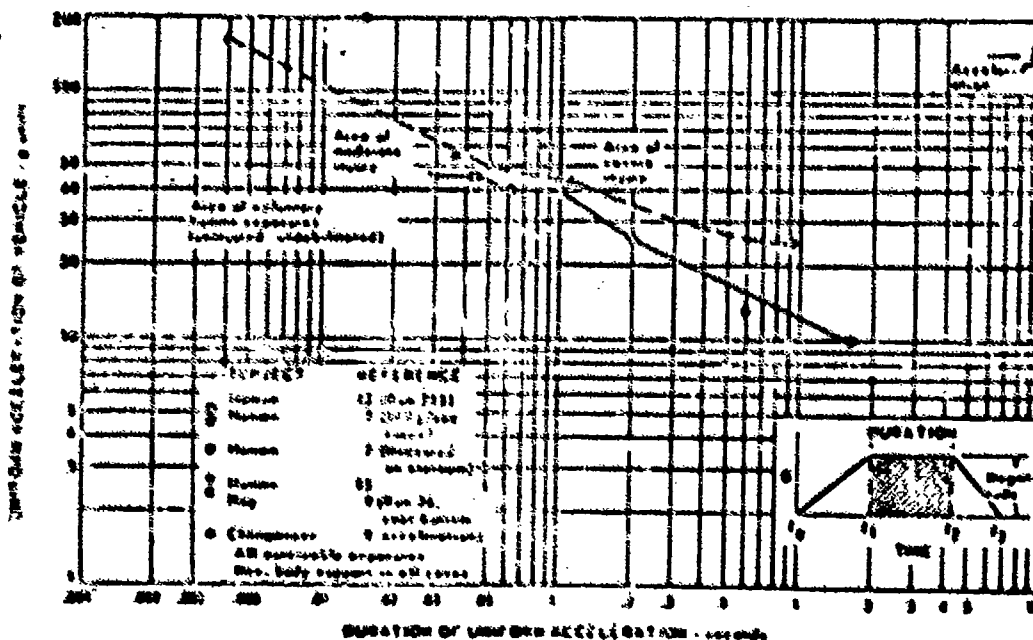


These two graphs show the durations and magnitudes of abrupt decelerations in the  $G_z$  (longitudinal) direction which have been endured by various animals and man, showing areas of voluntary endurance without injury, moderate injury, and severe injury marked by shading. Graph *a* shows data of  $+G_z$  acceleration (headward), and *b* shows data for  $-G_z$  acceleration (tailward). Reference numbers on the graphs are those in the original reports.

a.



b.



These two graphs show the durations and magnitudes of abrupt transverse decelerations which have been endured by various animals and man, showing areas of voluntary endurance without injury, moderate injury, and severe injury. (Graph a summarizes (+G<sub>y</sub>) data (chest to back acceleration) and b shows (-G<sub>y</sub>) data (back to chest acceleration). Reference numbers on the graphs are those in the original reports.

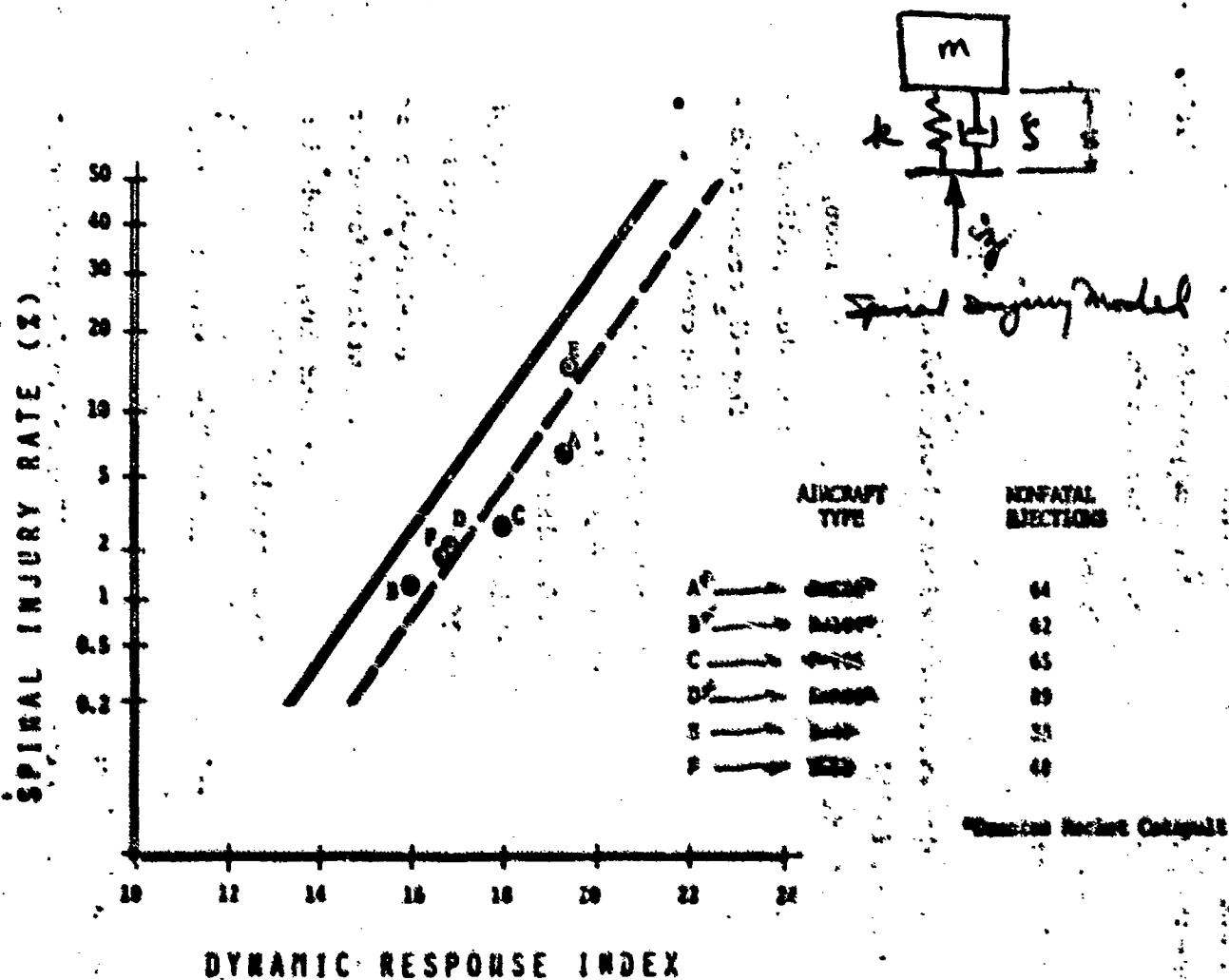


Figure 9.

PROBABILITY OF SPINAL INJURY ESTIMATED FROM LABORATORY DATA COMPARED TO OPERATIONAL EXPERIENCE

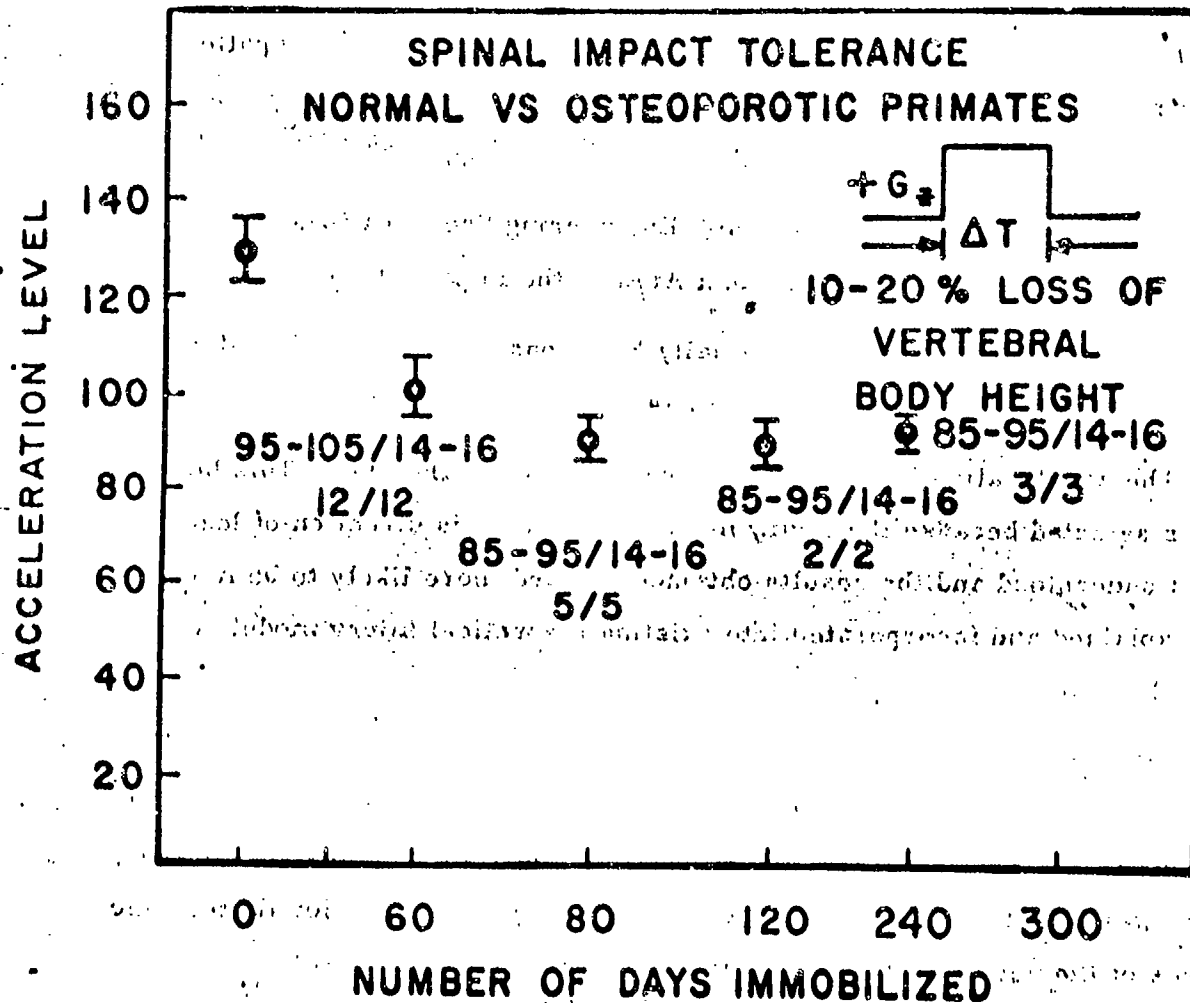


Figure 10. Spinal Impact Tolerance of the Normal and Osteoporotic Primates

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**Figure 11. Tolerance of Stiff-Legged Standing Men to Shock Motion of Short Duration**

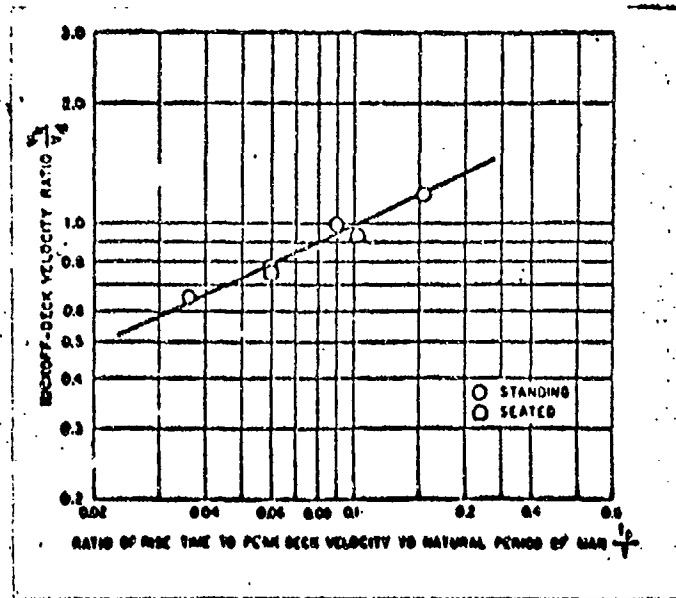
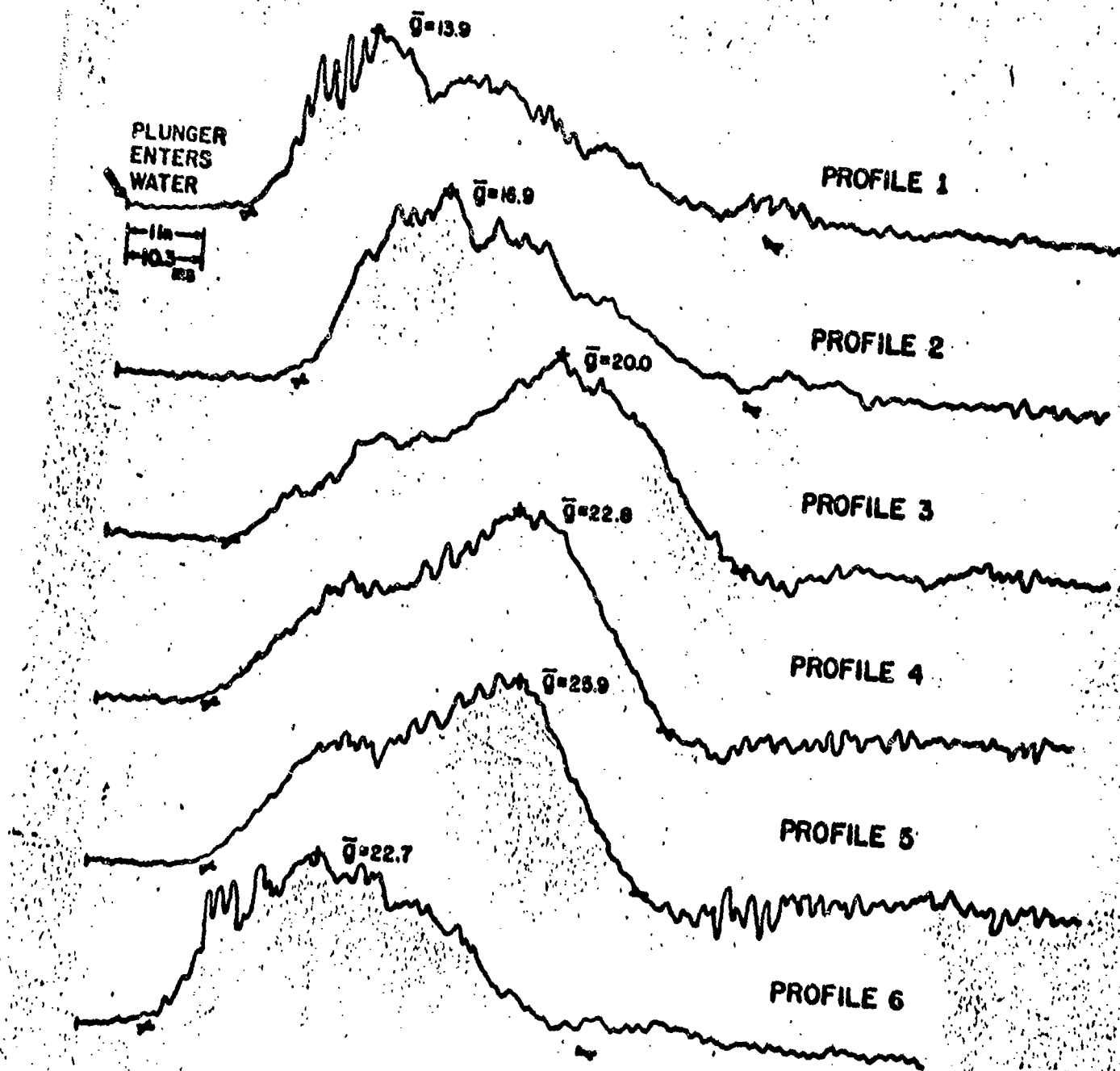
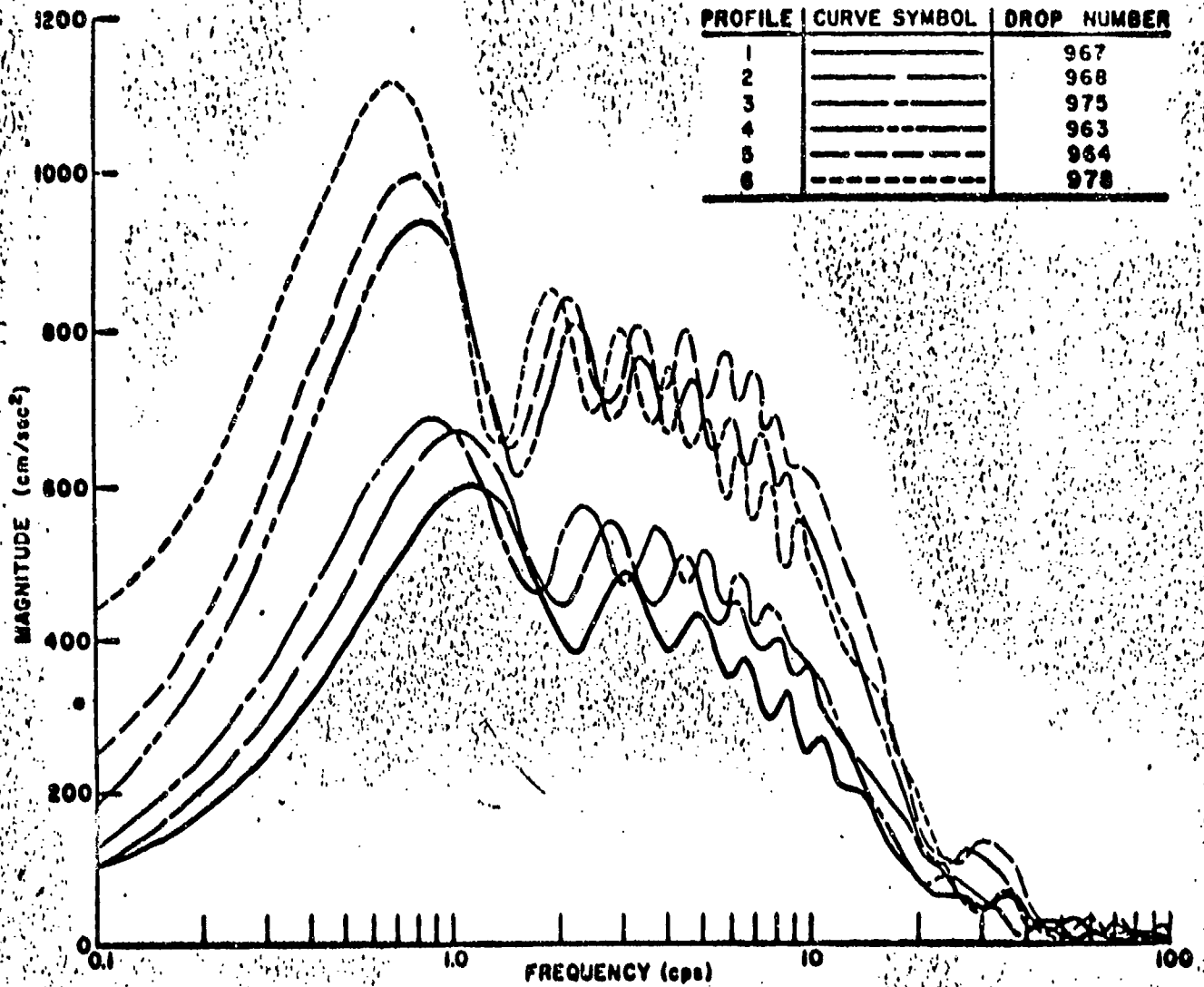


Figure 12. Ratio of Kickoff to Peak Deck Velocity as a Function of Ratio of Rise Time to Peak Deck Velocity to Natural Period of Man



13a  
Figure 2. ~~Vehicle Acceleration Profiles~~ Vehicle Acceleration Profiles

68



136  
Figure 2 Power Density Spectra



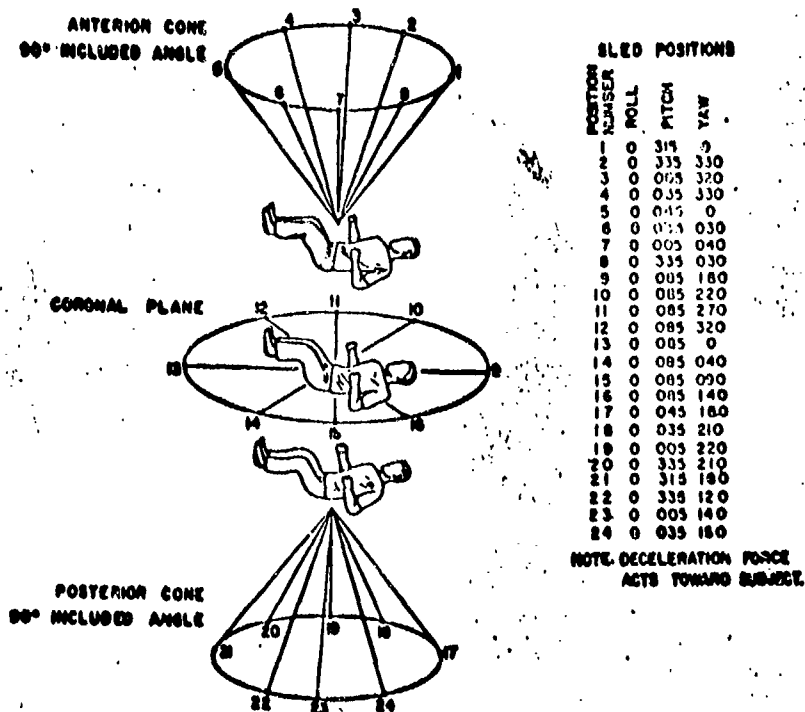
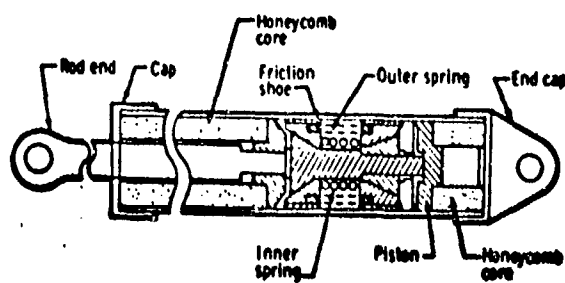
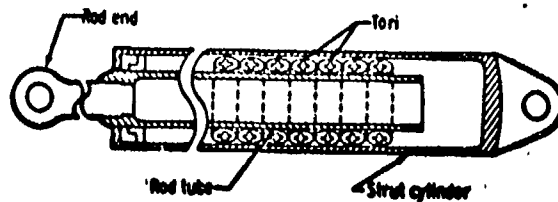


Figure 14. Deceleration Force Vector Orientation for Apollo Impact Tests



(a) Honeycomb strut.



(b) Cyclic strut.

Figure 15. Comparison of Honeycomb and Cyclic Strain Impact Attenuation Streets for Apollo